DYNAMIC SHIMSET CALIBRATION FOR B_0 OFFSET

DESCRIPTION

The following relates to the magnetic resonance arts. It finds particular application in magnetic resonance imaging, and will be described with particular reference thereto. However, it also finds application in magnetic resonance spectroscopy and other techniques that benefit from a main B_0 magnetic field of precisely known magnitude.

In magnetic resonance imaging, a temporally constant main B_0 magnetic field is produced that is spatially uniform at least over a field of view. Achieving sufficient uniformity for larger main B_0 magnetic field strengths, such as 3 Tesla or higher, can be difficult. Non-uniformities in the main B_0 magnetic field can produce various types of image artifacts. For example, in echo planar imaging, main field non-uniformities can lead to pixel shifting in the reconstructed images. Design tradeoffs to achieve hardware cost reduction, greater compactness of scanners, more open access for the subject or patient, and so forth also may contribute to magnetic field non-uniformities

Main B_0 magnetic field uniformity can be improved using active shimming, in which dedicated shim coils produce a supplementary or shim magnetic fields that compensate for non-uniformities of the magnetic field produced by the main magnet. The main magnet is usually superconducting, while the shim coils are usually resistive coils. In one embodiment, each shim coil produces a magnetic field having a spatial distribution that is functionally orthogonal to the magnetic fields produced by the other shim coils. For example, each shim coil can produce a magnetic field having a spatial distribution corresponding to Legendre polynomials or spherical harmonic components.

To calibrate the shim currents, a magnetic field probe or other device, or a dedicated magnetic resonance sequence executed by the scanner, is used to measure the spatial distribution of the main B_0 magnetic field without the shim coils energized. The spatial distribution is decomposed into orthogonal spatial components such as spherical harmonic terms. Orthogonal terms of the unshimmed magnetic field which should be increased are supplemented using corresponding shim coils, while orthogonal terms which should be decreased are partially canceled by energizing corresponding shim coils to produce opposing shim fields.

Typically, the shim currents are calibrated infrequently, such as when the magnetic resonance scanner is installed, after major maintenance, or the like. The stored shim current calibration values are applied during magnetic resonance imaging sessions to improve main B_0 field uniformity.

At higher main B_0 magnetic fields, such as at about 3 Tesla or higher, magnetic properties of the imaged subject, such as the magnetic susceptibility, increasingly distort the main B_0 magnetic field. These distortions are generally imaging subject-dependent, and may also depend upon the positioning of the imaging subject and the region of interest of the subject which is being imaged. In such situations, it becomes advantageous to perform dynamic shimming, in which shim coil currents are adjusted for each specific imaging subject, and perhaps are adjusted during an imaging session as the imaged region shifts.

To perform shimming that accounts for distortion caused by the imaging subject, the main B_0 magnetic field is measured with the imaging subject *in situ* using magnetic field sensors disposed in the magnet or a magnetic field mapping pulse sequence executed by the magnetic resonance imaging scanner. The mapped spatial distribution of the main B_0 magnetic field is decomposed into orthogonal components and suitable corrective shim coil magnetic fields are determined and applied.

Shim coils are designed to adjust the main B_0 magnetic field which is directed along a selected main field axis. In typical horizontal bore magnets, this axis typically lies along the bore axis and is designated as the z-axis; however, vertical magnets or other geometric configurations can also be employed. Hence, the shim coils are designed principally to produce a magnetic field component parallel to the main field axis (for example parallel to the z-axis for a horizontal bore magnet) to enable spatially selective enhancement or partial cancellation of the main B_0 magnetic field. However, the shim coils also produce some components transverse to the main field axis (for example perpendicular to the z-axis for a horizontal bore magnet).

These transverse shim magnetic field components contribute to a shift in the magnitude of the shimmed main B_0 magnetic field, and hence contribute to a shift in the resonance frequency. The shimming-induced magnetic field magnitude shift depends upon the magnitude of the shim currents applied. Such magnetic field magnitude shifts are problematic for imaging techniques that depend on having a precise main field. For example, in echo planar imaging, compact spiral k-space trajectory imaging, chemical shift

selective excitation, and some other techniques, the magnitude shift of the main field due to shimming can produce pixel shifting or other deleterious image artifacts.

The present invention contemplates an improved apparatus and method that overcomes the aforementioned limitations and others.

According to one aspect, a magnetic resonance imaging method is provided. A magnitude shift of a main B_0 magnetic field responsive to energizing one or more shim coils at selected shim currents is determined. The one or more shim coils are energized at the selected shim currents. A correction is performed during the energizing to correct for the determined magnitude shift of the main B_0 magnetic field.

According to another aspect, a magnetic resonance imaging apparatus is disclosed. A means is provided for generating a main B_0 magnetic field. One or more shim coils shim the main B_0 magnetic field. A means is provided for determining a magnitude shift of the main B_0 magnetic field responsive to energizing the one or more shim coils at selected shim currents. A means is provided for energizing the one or more shim coils at the selected shim currents. A means is provided for performing a correction during the energizing to correct for determined magnitude shift of the main B_0 magnetic field.

According to yet another aspect, a magnetic resonance imaging scanner is disclosed. A main magnet generates a main B_0 magnetic field. One or more shim coils selectively shim the main B_0 magnetic field at selected shim currents. A processor executes a process including determining a magnitude shift of the main B_0 magnetic field responsive to the selective shimming.

One advantage resides in facilitating patient-specific shimming.

Another advantage resides in facilitating dynamic shimming during imaging.

Yet another advantage resides in improved image quality due to a close agreement between the shimmed main B_0 magnetic field magnitude and tuning of the radio frequency transceiver.

Numerous additional advantages and benefits will become apparent to those of ordinary skill in the art upon reading the following detailed description of the preferred embodiments.

The invention may take form in various components and arrangements of components, and in various process operations and arrangements of process operations. The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 diagrammatically shows a magnetic resonance imaging system implementing patient-specific and/or dynamic main B_0 magnetic field shimming.

FIGURE 2 diagrammatically plots the typical effect of increased shimming on the magnetic resonance frequency distribution in the main B_0 magnetic field.

FIGURE 3 diagrammatically shows vector computation of the magnitude shift of the main B_0 magnetic field magnitude due to shimming.

FIGURE 4 diagrammatically shows dynamic shimming implemented by separately shimming four imaging regions of the volume of interest.

With reference to FIGURE 1, a magnetic resonance imaging scanner **10** includes a housing **12** defining a generally cylindrical scanner bore **14** inside of which an associated imaging subject **16** is disposed. Main magnetic field coils **20** are disposed inside the housing **12**, and produce a main B_0 magnetic field parallel to a central axis **22** of the scanner bore **14**. In FIGURE 1, the direction of the main B_0 magnetic field is parallel to the z-axis of the reference x-y-z Cartesian coordinate system. Main magnetic field coils **20** are typically superconducting coils disposed inside cryoshrouding **24**, although resistive main magnets can also be used.

The housing **12** also houses or supports magnetic field gradient coils **30** for selectively producing magnetic field gradients parallel to the central axis **22** of the bore **14**, along in-plane directions transverse to the central axis **22**, or along other selected directions. The housing **12** further houses or supports a radio frequency body coil **32** for selectively exciting and/or detecting magnetic resonances. An optional coil array **34** disposed inside the bore **14** includes a plurality of coils, specifically four coils in the illustrated example coil array **34**, although other numbers of coils can be used. The coil array **34** can be used as a phased array of receivers for parallel imaging, as a sensitivity encoding (SENSE) coil for SENSE imaging, or the like. In one embodiment, the coil array

34 is an array of surface coils disposed close to the imaging subject **16**. The housing **12** typically includes a cosmetic inner liner **36** defining the scanner bore **14**.

The coil array **34** can be used for receiving magnetic resonances that are excited by the whole body coil **32**, or the magnetic resonances can be both excited and received by a single coil, such as by the whole body coil **32**. It will be appreciated that if one of the coils **32**, **34** is used for both transmitting and receiving, then the other one of the coils **32**, **34** is optionally omitted.

The main magnetic field coils **20** produce a main B_0 magnetic field. A magnetic resonance imaging controller **40** operates magnetic field gradient controllers **42** to selectively energize the magnetic field gradient coils **30**, and operates a radio frequency transmitter **44** coupled to the radio frequency coil **32** as shown, or coupled to the coils array **34**, to selectively energize the radio frequency coil or coil array **32**, **34**. By selectively operating the magnetic field gradient coils **30** and the radio frequency coil **32** or coil array **34** magnetic resonance is generated and spatially encoded in at least a portion of a region of interest of the imaging subject **16**. By applying selected magnetic field gradients via the gradient coils **30**, a selected k-space trajectory is traversed, such as a Cartesian trajectory, a plurality of radial trajectories, or a spiral trajectory. Alternatively, imaging data can be acquired as projections along selected magnetic field gradient directions. During imaging data acquisition, a radio frequency receiver **46** coupled to the coils array **34**, as shown, or coupled to the whole body coil **32**, acquires magnetic resonance samples that are stored in a magnetic resonance data memory **50**.

The imaging data are reconstructed by a reconstruction processor **52** into an image representation. In the case of k-space sampling data, a Fourier transform-based reconstruction algorithm can be employed. Other reconstruction algorithms, such as a filtered backprojection-based reconstruction, can also be used depending upon the format of the acquired magnetic resonance imaging data. For SENSE imaging data, the reconstruction processor **52** reconstructs folded images from the imaging data acquired by each coil and combines the folded images along with coil sensitivity parameters to produce an unfolded reconstructed image.

The reconstructed image generated by the reconstruction processor **52** is stored in an image memory **54**, and can be displayed on a user interface **56**, stored in non-volatile memory, transmitted over a local intranet or the Internet, viewed, stored, manipulated, or

so forth. The user interface **56** can also enable a radiologist, technician, or other operator of the magnetic resonance imaging scanner **10** to communicate with the magnetic resonance imaging controller **40** to select, modify, and execute magnetic resonance imaging sequences.

The main magnetic field coils **20** generate the main B_0 magnetic field, preferably at about 3 Tesla or higher, which is substantially uniform in the imaging volume of the bore **14**. However, some non-uniformity may be present or may develop over time due to mechanical or electronic drift of components of the scanner **10**. The amount of image distortion caused by such non-uniformity may depend upon the location of imaging within the bore **14**. Moreover, when the associated imaging subject **16** is inserted into the bore **14**, the magnetic properties of the imaging subject can distort the main B_0 magnetic field.

To improve the uniformity of the main B_0 magnetic field, one or more shim coils **60** housed or supported by the housing **12** provide active shimming of the main B_0 magnetic field. In one embodiment, each shim coil produces a shimming magnetic field having a spatial distribution that is functionally orthogonal to the magnetic fields produced by the other shim coils. For example, each shim coil can produce a magnetic field having a spatial distribution corresponding to a spherical harmonic component. By selectively energizing the various shim coils **60** at selected shim currents, non-uniformities of the main B_0 magnetic field are reduced.

Ideally, each shim coil produces a magnetic field distribution within the bore **14** that includes only B_z components, that is, magnetic fields directed parallel to the main B_0 magnetic field parallel to the z-direction, with no transverse B_x or B_y components. The B_z components are selected to enhance or partially cancel the main B_0 magnetic field produced by the main magnetic field coils **20** to correct for inherent non-uniformities, for distortion caused by the imaging subject **16**, or the like. Specifically, a shim currents processor **62** determines appropriate shim currents for one or more of the shim coils **60** to reduce non-uniformity of the main B_0 magnetic field. The shim currents processor **62** selects appropriate shim currents based on known configurations of the shim coils **60** and based on information on the magnetic field non-uniformity that needs to be corrected. Non-uniformity of the main B_0 magnetic field can be determined in various ways, such as by acquiring a magnetic field map using a magnetic field mapping magnetic resonance sequence executed by the scanner **10**, by reading optional magnetic field sensors (not

shown) disposed in the bore **14**, by performing *a priori* computation of the expected magnetic field distortion produced by introduction of the imaging subject **16**, or so forth. Magnetic field measurement sequences may be intermixed with the imaging sequence to check the main B_0 magnetic field magnitude periodically, e.g. after each slice. The shim currents processor **62** controls a shims controller **64** to energize one or more of the shim coils **60** at the selected shim currents.

Although it would be desirable for each shim coil to produce a magnetic field distribution within the bore **14** that includes only B_z components, because the magnetic flux must follow a closed loop, the shim coils **60** typically also produce at least some residual transverse magnetic field components, such as B_x and/or B_y components, in at least a portion of the bore **14**. A consequence of these transverse magnetic field components is that while the shimming reduces spatial non-uniformity of the main B_0 magnetic field, the average or mean magnitude $|B|$ of the main B_0 magnetic field changes, and usually increases, with increased shimming.

The resonance frequency f_{res} at a given point in space is given by:

$$f_{res} = \gamma|B(x,y,z)| \qquad (1),$$

where $|B(x,y,z)|$ is the magnitude of the magnetic field at position (x,y,z) and γ is the gyrometric ratio for the excited nuclear magnetic resonance. The magnitude $|B(x,y,z)|$ depends upon the total field, not just the B_z component. Using the Cartesian coordinate system designated in FIGURE 1:

$$|B(x,y,z)|^2 = [B_x(x,y,z)]^2 + [B_y(x,y,z)]^2 + [B_z(x,y,z)]^2 \qquad (2).$$

As an example, ^{1}H proton nuclei have a gyrometric ratio γ=42.58 MHz/Tesla, so at $|B(x,y,z)|$=3.0 Tesla the resonance frequency is approximately f_{res}=128 MHz. Equation (1) indicates that the frequency distribution of magnetic resonance intensity thus corresponds to the distribution of the magnitude of the magnetic field in the imaging volume.

With reference to FIGURE 2, which plots the distribution of magnetic resonance intensity as a function of frequency, the observed effect of shimming on the magnitude of the main B_0 magnetic field is illustrated. In FIGURE 2, the unshimmed magnetic

resonance intensity distribution as a function of frequency, denoted $I_0(f)$, is relatively broad and centered at an unshimmed center frequency f_0. The breadth of the unshimmed magnetic resonance intensity distribution $I_0(f)$ reflects a substantial spatial non-uniformity of the unshimmed main B_0 magnetic field in the bore **14**. As shimming is applied using shimming currents selected to reduce the field non-uniformity, the magnetic resonance intensity distribution becomes narrower, reflecting improved spatial uniformity. In FIGURE 2, a shimmed, substantially spatially uniform magnetic field provides a narrow magnetic resonance intensity distribution denoted $I_s(f)$.

In addition to being substantially narrowed, however, the shimmed magnetic resonance intensity distribution $I_s(f)$ is also shifted toward higher frequency, and has a center frequency $f_s>f_0$. For applications in which the shimming may be adjusted relatively frequently, this shift in the resonance frequency can be problematic and can lead to image artifacts. In applications where dynamic shimming is performed during imaging, such frequency shifting occurs during the imaging.

With reference to FIGURE 3, a vector computation of the magnitude shift of the main B_0 magnetic field magnitude due to shimming is drawn. The desired shimmed magnetic field has a component in the z-direction of magnitude B_z. If at a given position (x,y,z) the magnetic field produced by the main magnetic field coils **20** is less than B_z, then the shimming preferably enhances that field to the value B_z. Similarly, if the magnetic field produced by the main magnetic field coils **20** is greater than B_z, then the shimming preferably partially cancels that field to match the value B_z.

Thus, the shimmed magnetic field has a substantially spatially uniform B_z component indicated in FIGURE 3 throughout the bore **14**. However, any additional, undesired transverse magnetic field components produced by the shimming, such as the illustrated component $B_x(+I_1)$ or the illustrated component $B_x(-I_2)$, both oriented along the x-direction, are not accounted for in determining the desired shim current. Thus, as illustrated in FIGURE 3, if a shim current $+I_1$ is required to produce the B_z field, and this shim current $+I_1$ produces an additional undesired transverse component $B_x(+I_1)$, then the total field at the position (x,y,z) is $|B|(+I_1)=(B_z^2 + [B_x(+I_1)]^2)^{0.5}$ which is greater than the desired magnitude B_z. Similarly, if a shim current $-I_2$ is required to produce the B_z field, and this shim current $-I_2$ produces an additional undesired transverse component $B_x(-I_2)$, then the total field at the position (x,y,z) is $|B|(-I_2)=(B_z^2 + [B_x(-I_2)]^2)^{0.5}$ which is also greater

than the desired magnitude B_z. Indeed, it will be appreciated that any transverse component, regardless of its positive or negative sense or its transverse orientation, will tend to increase the magnitude of the shimmed magnetic field. These effects of undesired transverse components are typically small, since the shim components are typically smaller than the main B_0 component, and the nature of the vector magnitude operation depends only weakly on spatially orthogonal components which are smaller than the largest component. However, the requirement that the magnetic field flux lines form a closed loop typically prevents the transverse components from being identically zero everywhere within the bore **14**.

These transverse magnetic fields, and their contribution to the total vector magnitude magnetic field are referred to here as Maxwell terms. In some literature, they are also sometimes referred to as Maxwell fields or concomitant fields.

With reference returning to FIGURE 1, a magnitude shift processor **70** determines the magnitude shift of the main B_0 magnetic field expected to occur responsive to energizing one or more of the shim coils **60** at the shim currents selected by the shim currents processor **62**. The magnitude shift processor **70** performs this computation before the shim coils **60** are actually energized, to provide an *a priori* prediction of the magnitude shift. The *a priori* computation can be performed by accessing a previously determined magnitude shift calibration table **72** that stores magnitudes shifts previously measured for various shim currents and combinations of shim currents. For example, the magnetic resonance intensity distribution can be measured as a function of frequency for various shim currents and combinations of shim currents to determine shifted frequencies f_s for the various shim currents and current combinations as illustrated in FIGURE 2. Based on Equation (1), the magnitude shift $\Delta|B_0|$ of the main B_0 magnetic field can be computed as:

$$\Delta|B_0| = |B_0|_{shimmed} - |B_0|_{unshimmed} = (f_{shimmed} - f_{unshimmed})/\gamma \quad (3),$$

where γ is again the gyrometric ratio for the measured nuclear magnetic resonances. While this empirical approach is straightforward, it generally requires measuring a large number of combinations of shim currents. Moreover, if the selected combination of shim currents

is not included in the calibration table **72**, potentially computationally intensive numerical interpolation is typically employed.

In another approach, magnitude shift of the main B_0 magnetic field is estimated using Maxwell terms. This approach recognizes that since the shim coils **60** are intended to produce magnetic fields oriented in the z-direction, the inequality $B_z >> B_x, B_y$ typically holds. That is, the field component along the z-direction is typically much larger than the magnetic field components transverse to the z-direction. Under this condition, the magnitude shift $\Delta|B_0|$ can be represented as:

$$\Delta|B_0| \cong [^{1}K_s][I_s] + [^{2}K_s][I_s^{2}] + [^{4}K_s][I_s^{4}] + \ldots + [^{2n}K_s][I_s^{2n}] \quad (4),$$

where $[I_s]$ is a vector of shim currents applied to the shims **60**. A zero element of the vector $[I_s]$ indicates that the corresponding shim is not energized and thus does not contribute to the magnitude shift $\Delta|B_0|$. The coefficients vector $[^{1}K_s]$ is a zeroeth order coefficients vector of calibrated coefficients for the shims **60**, and describes the direct B_0 term created by each of the shims **60**. The coefficients vector $[^{2}K_s]$ is a first order Maxwell term coefficients vector of calibrated coefficients for the shim coils **60** that describes the first Maxwell term contribution created by each of the shim coils **60**. The vector $[I_s^{2}]$ is a vector containing the shim current-squared values of shim currents applied to the shims **60**. Again, a zero element in the vector $[I_s^{2}]$ indicates that the corresponding shim is not energized and thus does not contribute to the magnitude shift $\Delta|B_0|$. Similarly, the coefficients vectors $[^{4}K_s] \ldots [^{2n}K_s]$ represent the 2nd through nth Maxwell term coefficients, and the vectors $[I_s^{4}] \ldots [I_s^{2n}]$ represent vectors of the shim current values raised to the indicated powers.

The Maxwell coefficients vectors $[^{\cdot\cdot}K_s]$ are stored in a Maxwell coefficients vectors memory **74**. In one embodiment, these coefficients are calibrated by measuring the magnetic field shift $\Delta|B_0|$ with each shim energized separately at one or a few shim current levels. The elements of the Maxwell coefficients vectors $[^{\cdot\cdot}K_s]$ for that shim coil are calibrated by optimizing the coefficients for that shim coil using Equation (4) with the $[I_s^{n}]$ vectors having zero elements except for elements corresponding to the energized shim. This calibration assumes that the magnitude shifts of the individually operated shim coils

additively combine when two or more of the shim coils **60** are operated together, which is a convenient simplifying assumption.

Advantageously, once the Maxwell coefficients $[\ddot{K}_s]$ are calibrated for the shim coils **60**, the magnitude shift of the main B_0 magnetic field can be computed *a priori* for substantially any combination of selected shim currents, even combinations other than those used in the calibration, by evaluating Equation (4) using the selected shim currents as input values. The empirical functional relationship is provided in Equation (4) is a continuous function with respect to the shim currents, as compared with the discrete values typically stored in the calibration table **72**, and so potentially computationally intensive numerical interpolation is generally not employed.

Instead of empirically calibrating the Maxwell coefficients $[\ddot{K}_s]$, these coefficients can be computed from first principles based on the geometric configurations of the shim coils **60**. Such first principles computations can be performed, for example, using finite element modeling of the coil geometries for various simulated shim currents and fitting the coefficients to the simulation results.

The magnitude shift $\Delta|B_0|$ of the main B_0 magnetic field computed by the magnitude shift processor **70** is used to perform a correction during the energizing of the selected one or more of the shim coils **60** to correct for the determined magnitude shift of the main B_0 magnetic field. In one embodiment, the magnitude shift $\Delta|B_0|$ computed by the magnitude shift processor **70** is compensated by operating a D.C. shim controller **80** to energize a D.C. shim coil **82**. The D.C. shim coil **82** is a zero order shim coil that when energized produces a spatially uniform magnetic field in the bore **14**. The D.C. shim controller **80** energizes the D.C. shim **82** at a shim current selected to oppose and substantially cancel the magnitude shift $\Delta|B_0|$ (assuming a positive magnitude shift). The D.C. shim **82** cancels the positive magnitude shift $\Delta|B_0|$ to maintain the main B_0 magnetic field at a constant value even when one or more of the shims **60** are operating.

In another embodiment, the magnitude shift processor **70** outputs a magnetic resonance frequency shift Δf_{res} equivalent to the magnitude shift $\Delta|B_0|$ of the main B_0 magnetic field. As shown by Equation (1), the magnetic resonance frequency shift Δf_{res} is equal to the magnitude shift $\Delta|B_0|$ except for the scaling gyrometric ratio factor γ. The magnetic resonance frequency shift Δf_{res} output by the magnitude shift processor **70** is used as control signals (indicated by dashed connecting arrows in FIGURE 1) to control the

radio frequency transceiver **44**, **46** including the radio frequency transmitter **44** and the radio frequency receiver **46** to ensure that they are operating at the magnetic resonance frequency corresponding to the main B_0 magnetic field including the magnitude shift $\Delta|B_0|$. In other words, with reference to FIGURE 2 the center frequency of the transmitter **44** is tuned to the shimmed frequency f_s. An analogous adjustment can be made at the receiver **46**.

Any of the above-described magnitude shift correction embodiments or their equivalents can be employed to facilitate adjusting the shimming on a relatively frequent basis. For example, shimming can be adjusted for each patient, to account for different magnetic susceptibility properties of each patient. Moreover, any of the described magnitude shift correction embodiments or their equivalents facilitate dynamic shimming during imaging, in which the shimming is adjusted on a regional, per-slice, or other basis during the imaging session of a single patient.

With reference to FIGURE 4, an imaging volume **V** encompasses the head and torso of the imaging subject **16**. The unshimmed main B_0 magnetic field is distorted in a spatially non-uniform fashion across the imaging volume **V** by the imaging subject **16**. In FIGURE 4, this distortion is diagrammatically represented by plotting the unshimmed average main B_0 magnetic field $|B(z)|$, averaged over each axial slice, as a function of axial slice position in the z-direction. While the variation in the z-direction is plotted, it will be recognized that the main B_0 magnetic field may be distorted in the transverse x and y directions as well. The entire imaging volume **V** could be shimmed as a unit; however, imposing spatial uniformity on the large volume **V** may be difficult.

In the dynamic shimming approach illustrated in FIGURE 4, the imaging volume **V** is divided up into four regions $\mathbf{R_1}$, $\mathbf{R_2}$, $\mathbf{R_3}$, $\mathbf{R_4}$ along the z-direction. Some regions exhibit more magnetic field variation than others. In the illustrated example, the regions $\mathbf{R_3}$, $\mathbf{R_4}$ have more magnetic field variation than the regions $\mathbf{R_1}$, $\mathbf{R_2}$. Each region $\mathbf{R_1}$, $\mathbf{R_2}$, $\mathbf{R_3}$, $\mathbf{R_4}$ is separately shimmed. That is, for each region, one or more shim currents are selected to substantially reduce non-uniformity of the main B_0 magnetic field in that region. Because the shimming is focused on smaller regions, more accurate shimming of each region can be performed. When the region $\mathbf{R_1}$ is imaged, the shim currents selected to shim that region are employed. When the region $\mathbf{R_2}$ is imaged, the shim currents selected to shim that region are employed. When the region $\mathbf{R_3}$ is imaged, the shim currents selected to shim

that region are employed. When the region $\mathbf{R_4}$ is imaged, the shim currents selected to shim that region are employed.

FIGURE 4 also diagrammatically plots the shimmed average main B_0 magnetic field $|B(z)|$ averaged over each axial slice during imaging of that slice. The shimmed average main B_0 magnetic field in each region is substantially uniform, but exhibits a magnitude shift $\Delta|B_0|$, which is not corrected in FIGURE 4. The plots of the shimmed average main B_0 magnetic fields $|B(z)|$ do not include optional compensation via the D.C. shim coil **82**. Because each region $\mathbf{R_1}$, $\mathbf{R_2}$, $\mathbf{R_3}$, $\mathbf{R_4}$ is imaged using generally different selected shim currents, the size of the magnitude shift $\Delta|B_0|$ differs for each of the four regions. In the illustrated example, larger shim currents are applied during imaging of regions $\mathbf{R_3}$, $\mathbf{R_4}$ to compensate for the relatively large magnetic field non-uniformities in those regions prior to shimming; whereas, smaller shim currents are applied during imaging of regions $\mathbf{R_1}$, $\mathbf{R_1}$ which exhibit less field non-uniformity. Correspondingly, the shimmed main B_0 magnetic field in regions $\mathbf{R_3}$, $\mathbf{R_4}$ have larger magnitude shifts $\Delta|B_0|$ as compared with regions $\mathbf{R_1}$, $\mathbf{R_1}$. By using the magnitude shift processor **70** to compute the magnitude shift $\Delta|B_0|$ appropriate for each selected shim currents combination used to shim each respective region $\mathbf{R_1}$, $\mathbf{R_2}$, $\mathbf{R_3}$, $\mathbf{R_4}$, the changing magnitude shifts $\Delta|B_0|$ during the dynamically shimmed imaging is compensated.

While FIGURE 4 illustrates four regions each including a plurality of slices, it will be appreciated that the dynamic shimming technique could be applied to other sub-volumes. For example, the dynamic shimming can be applied on a per-slice basis, in which shim currents are selected for each axial slice prior to imaging that slice.

In the embodiments heretofore described, shim currents are selected to reduce non-uniformity of the main B_0 magnetic field in an imaging region. Once the shim currents are selected, the magnitude shift $\Delta|B_0|$ produced by those selected shim currents is computed, and a correction for that computed magnitude shift $\Delta|B_0|$ is performed. The processes of selecting shim currents, computing the magnitude shift, and correcting are performed separately.

However, in other contemplated embodiments the processes of selecting shim currents, computing $\Delta|B_0|$, and correcting are partially or wholly integrated together. For example, the shim currents can be determined by optimizing a figure of merit that includes a field uniformity component and a magnitude shift component $\Delta|B_0|$. In this embodiment,

the shim currents, including shim currents for the shim coils **60** and the D.C. shim coil **82**, are simultaneously optimized by minimizing or maximizing the figure of merit, thus simultaneously performing the selecting of the shim currents and the computing of a correction of the magnitude shift $\Delta|B_0|$.

Shimming affects volumes, and measurement of resonant frequency occurs over volumes, these volumes typically exhibiting spatial dependences. In one embodiment, shifts may be measured as an average over a predefined volume, for example, a 20 centimeter diameter spherical reference volume located at the center of the magnet. Other contemplated embodiments may include volume definitions such as (i) the extent of a planned subsequent imaging region, (ii) some fraction of the central region of a prescribed imaging volume, (iii) the physical extent of the subject to be imaged, perhaps limited within a larger predefined volume, (iv) a typical volume defined depending upon the human anatomy of interest, or (v) a region explicitly defined by the operator performing the MRI procedure. Numerous other definitions are possible.

Furthermore, the determination of the resonance frequency shift induced by an energized shim may incorporate the choice of the volume in any of several ways. Magnetic field shifts or shift coefficients may be defined for one or more predefined volumes. Shifts may be characterized with spatial dependences, such as by fitting polynomials or other spatial functions. Such polynomials may be spherical harmonics, or they may match the spatial distributions of the respective Maxwell terms of each shim coil, for example. Shifts or shift coefficients may be determined at each of several points in discretized maps, and stored as volume representations.

For purposes of illustration, a specific embodiment of computation of Maxwell terms is now further described. Utilizing a spherical harmonic expansion of the magnetic fields, the main magnet coils **20** can be utilized mainly to generate the zeroeth order spherical harmonic of B_z. The magnetic field gradient coils **30** can be utilized to generate or to correct the first order spherical harmonic terms of B_z. The magnetic shim coils **60** can be utilized to generate or to correct the second order spherical harmonic terms of B_z. These second order shims may be referred to, for example, as $(x^2 - y^2)$, xy, xz, yz, and z^2. The spatial dependence of each of these shims matches its name, except for the shim named z^2, which generates a field with the spatial dependence function of $B_z = (z^2 - 0.5*(x^2 + y^2))$. For each of these second order shims, the corresponding transverse field can be

determined. The transverse fields B_x and B_y for each shim are constrained to a family of solutions, since the magnetic field must satisfy Maxwell's equations. Some freedom exists in the possible choice of the B_x and B_y functions.

In embodiments in which the shim coils are mechanically built on cylindrical surfaces, some symmetries can be incorporated into the solution. By imposing these symmetry constraints upon the candidate transverse fields B_x and B_y functions, the spatial dependence of each function is determined. For the $(x^2 - y^2)$ shim, the solution under these symmetry constraints is $B_x = 2xz$ and $B_y = -2yz$. For the xy shim, the solution under these symmetry constraints is $B_x = 2yz$ and $B_y = 2xz$. For the z^2 shim, the solution under these symmetry constraints is $B_x = -xz$ and $B_y = -yz$. Similar determinations can be performed for other second order shims.

In one embodiment, the total magnitude B shift is calculated for a shim current in any given shim coil by utilizing Equation (2) and integrating over a volume. The power series expansion of the square root function then yields coefficients for powers of the shim current. Only even powers will yield nonzero coefficients.

In another embodiment, the vector fields $\mathbf{B} = (B_x, B_y, B_{z)}$ for each shim are scaled proportional to the desired setting of the B_z component. The vectors for all the scaled shims are added. The magnitude of the summed vector is determined as a function of position x, y, and z.. The resultant function is integrated over a volume of interest to give a final shifted B magnitude.

Extension of the previous embodiment for successively higher orders of shims is straightforward and is readily performed by those skilled in the art based on the foregoing description relating to second order shims.

It is to be appreciated that correction of the resonance frequency for deviations induced by shims may be useful regardless of the mechanism which initially causes the change of frequency. Thus, an observed frequency shift might be induced through other mechanisms, besides the Maxwell terms described above. Mechanical deflections due to static magnetic forces, or thermal effects associated with currents in resistive shim coils are other examples of mechanisms which might induced frequency changes. These and other mechanisms may exhibit the same basic functional dependence, even though they may vary in strength significantly from the predicted Maxwell terms effect. For example, energizing a shim of a coil of a specific spherical harmonic might induce slight geometric

deflections in other magnetic structures, in turn inducing a frequency shift substantially proportional to the square of the shim current. Correction for such empirically observed magnetic field non-uniformities is readily performed using the foregoing calibration apparatuses and methods and straightforward variations thereof. Thus, the forgoing calibration apparatuses and methods are readily adapted to correct for empirically observed magnetic field non-uniformities observed or measured without identifying the underlying cause.

The invention has been described with reference to the preferred embodiments. Obviously, modifications and alterations will occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.